

1 UNEVEN-COUNTER-ROTATIONAL COIL BASED  
2 MRI RF COIL ARRAY

3 CROSS-REFERENCE TO RELATED APPLICATION

4 This application claims the benefit of U.S. provisional  
5 patent application Serial No. 60/273,092 filed March 2, 2001.

6 BACKGROUND OF THE INVENTION

7 Magnetic resonance imaging (MRI) relies on the detection of  
8 the MR signal from abundant protons in the human body. A radio  
9 frequency (RF) receive coil is a device to effectively "pick up"  
10 the MR signal from the background of noise for image production.  
11 MR signals induced in a RF receive coil are weak signals due to  
12 the very small population difference between the two relevant  
13 proton energy states at room temperature. One of the challenges  
14 in RF coil design is to improve the MR signal detection  
15 sensitivity.

16 One of the approaches to improve signal detection  
17 sensitivity and/or field of view is to use multiple receive coils  
18 as an array. The basic idea is that instead of making a larger  
19 and less sensitive coil that covers the entire volume of  
20 interest, plural smaller and more sensitive coils are distributed  
21 over the volume of interest. Each individual coil picks up  
22 signal and noise from a localized volume. With separate  
23 detection circuitry, each coil element receives the image signal  
24 simultaneously. Signals from all the coils are finally combined  
25 and processed to reconstruct the MR image for the entire volume  
26 of interest.

27 The principle of MRI involves exciting protons and detecting  
28 the resulting free induction decay signals. Each proton  
29 possesses a tiny magnetic moment precessing about the static

1 magnetic field. The macroscopic behavior of millions of protons  
2 can be represented by a resultant magnetization vector aligning  
3 with the static magnetic field  $B_0$ . A strong RF excitation pulse  
4 effectively tips the magnetization away from  $B_0$ . The free  
5 induction decay of this magnetization is detected in a plane  
6 perpendicular to  $B_0$ . Thus, for maximal signal induction, the  
7 normal direction of a receive coil must be perpendicular to the  
8 direction of the static magnetic field  $B_0$ .

9 Based on the direction of static magnetic field, commercial  
10 MRI systems are either horizontal or vertical. The so-called  
11 co-planar type coil arrays have proved to be effective for  
12 horizontal MRI systems for the reasons discussed in the previous  
13 paragraph. In a co-planar array, surface coils are arranged in a  
14 co-planar fashion and distributed over a volume of interest.

15 In general, such co-planar type surface coil arrays are not  
16 very effective for a vertical system because the condition  
17 required for maximal signal induction can hardly be fulfilled.  
18 Various modifications to the co-planar designs have been proposed  
19 with limited success.

20 It is known that solenoidal type coils have several  
21 advantages for a vertical field system, including its  
22 sensitivity, uniformity and its natural fit to various body  
23 parts. It is advantageous to utilize solenoidal based coil  
24 arrays for vertical MRI systems.

25 To successfully implement a solenoidal coil array, one must  
26 be able to isolate solenoidal coils of the array to prevent them  
27 from coupling to each other. This is required because all coils  
28 in a coil array typically receive signals simultaneously.  
29 "Cross-talk" between different coils is undesirable. Thus  
30 effective coil isolation is a major challenge in solenoidal coil  
31 array design.

32 A so-called sandwiched solenoidal array coil (SSAC) has been

1 set forth in U.S. Patent Application No. 09/408,506. A SSAC  
2 consists of two solenoidal receive coils, a counter-rotational  
3 solenoidal coil and a second solenoidal coil sandwiched between  
4 the two counter-rotational winding sections of the first coil.

5 The counter-rotational solenoidal coil produces a gradient  
6 B<sub>1</sub> field that has a double-peak "M" shape sensitivity profile.  
7 The second solenoidal coil produces a single-peak profile  
8 sandwiched between the two peaks of the "M" shape profile of the  
9 first coil.

10 The sensitivity profile of a SSAC is determined by the  
11 summation of an "M" shape double-peak profile and a centralized  
12 single-peak profile generated by the two coils. To avoid  
13 unwanted dark band artifacts in the array coil sensitivity  
14 profile, the geometric parameters of both coils must be set  
15 properly. This process is sensitive to the geometries at hand.

#### 16 SUMMARY OF THE INVENTION

17 A MRI RF coil array is formed from a first coil having a  
18 null B<sub>1</sub> point and a quasi-one-peak sensitivity profile, and a  
19 second coil oriented with respect to the first coil to reduce  
20 coupling.

#### 21 BRIEF DESCRIPTION OF THE DRAWINGS

22 FIG. 1 is a schematic diagram of an uneven-counter-  
23 rotational solenoidal coil.

24 FIG. 2 is a graphical diagram of exemplary B<sub>1</sub> and  
25 sensitivity profiles of an uneven-counter-rotational solenoidal  
26 coil according to the invention.

27 FIG. 3 is a graphical diagram of an exemplary sensitivity  
28 profile of a coil array according to the invention as a

1 superposition of two individual solenoidal coils.

2 FIG. 4 is a schematic diagram of a coil array according to  
3 the invention in a cascade configuration.

4 FIG. 5 is a schematic diagram of a coil array according to  
5 the invention in an overlapped configuration.

6 FIG. 6 is a schematic diagram of a coil array according to  
7 the invention in a sandwiched configuration.

8 FIG. 7 is a schematic diagram of an embodiment of the  
9 invention showing spacing parameters.

10 FIG. 8 is a graphical diagram of an exemplary coupling  
11 sensitivity between the two coils of the array of FIG. 7.

12 FIG. 9 is a graphical diagram of a B1 profile for coil array  
13 of FIG. 7.

14 DESCRIPTION OF THE PREFERRED EMBODIMENTS

15 Referring to FIG. 1, an uneven-counter-rotational (UCR) coil  
16 12 is illustrated. The coil 12 is formed from a first coil  
17 section A and a second coil section B. Section A has more turns  
18 than section B, for example, 3 versus 1. Section B is wound in  
19 the opposite direction from section A. For example, section A  
20 has three turns with the current flowing in the same direction  
21 and section B has one turn with current flowing in a counter-  
22 rotational direction. The separation between the neighboring  
23 turns is denoted as S12, S23 and S34, respectively. In general,  
24 the turn separation and diameter parameters may have different  
25 values depending on the specific coil design needs.

26 For example, the parameters may be as follows: S12=8cm,  
27 S23=7cm, S34=10cm and D=26.7cm. FIGS. 2a and 2b show the B1  
28 field produced by the sections A and B, respectively. FIG. 2c  
29 shows the total B1 field produced by the UCR solenoidal coil 12.  
30 The sensitivity profile is shown in FIG. 2d. It can be seen from

1 FIG. 2d that the UCR coil 12 generates a null-B1 point near  
2 location N and a quasi-one-peak sensitivity profile.

3 Referring to FIGS. 4, 5 and 6, a second solenoidal coil 14  
4 may be placed near the null-B1 point to form a solenoidal array  
5 with the UCR coil 12 while achieving good isolation between the  
6 two solenoidal coils 12, 14. In practice, an additional isolation  
7 capacitor may be used for the convenience of fine isolation  
8 adjustment if needed.

9 The second solenoidal coil 14 may, for example, be formed  
10 of multiple turns as needed. The number of turns and the  
11 separation between neighboring turns can be chosen to give a  
12 desired sensitivity profile and B1 strength. The corresponding  
13 sensitivity profile of the coil 14 partially overlaps with the  
14 profile of the UCR coil 12 to determine the sensitivity of the  
15 solenoidal coil array. FIG. 3 shows an example of a solenoidal  
16 coil array profile as the summation of the two solenoidal coils  
17 12, 14. FIG. 3 shows an artifact free array profile and the  
18 advantage of a quasi-one-peak UCR sensitivity profile design. A  
19 quasi one-peak profile for the UCR solenoidal coil 12 can be  
20 achieved by intentionally making the two peaks in the typical "M"  
21 shape profile uneven, i.e. the B1 field produced by one winding  
22 section of the UCR coil element is much stronger than the other.  
23 At the same time, the null-B1 point is retained in the  
24 quasi-one-peak profile, which is the basis for the inherent  
25 decoupling of the two solenoidal coils 12, 14. This can be  
26 accomplished by properly choosing the number of turns, their  
27 diameters and locations for each of the two winding sections.

28 A better understanding of the uneven-counter-rotational  
29 design, its quasi-one-peak profile and coil isolation between the  
30 two solenoidal coils of the array can be achieved by a closer  
31 look from the electromagnetic field point of view. First, the  
32 three turns in section A of the UCR coil 12 generate a strong B1

1 field as shown in FIG. 2a. The B1 field decreases gradually  
2 along the axis away from the section center. In fact, it  
3 approaches zero B1 at infinite distance from the center. If one  
4 would introduce a second solenoidal coil in a short distance from  
5 the section center, one would encounter strong coupling between  
6 the two coils.

7 Section B generates a B1 field of opposite direction to that  
8 of section A. Section B generates a negative B1 field of smaller  
9 peak value and different profile shape than that by section A.  
10 At certain location, the B1 field generated by sections A and B  
11 may cancel, forming a null-B1 point in the combined B1 profile of  
12 this UCR coil as shown in Figure 2c. By definition, the  
13 solenoidal coil 14 introduced to the location where the B1 field  
14 generated by sections A and B of the UCR coil 12 cancel  
15 experiences no magnetic coupling with the UCR coil 12. The  
16 null-B1 point can be set to be outside the UCR sections A, B,  
17 between the two UCR sections A, B or overlapped with one of the  
18 UCR sections.

19 The B1 field generated by the counter-rotational section B  
20 may cancel that by element A at different locations along the  
21 axis depending on relative field strength. Accordingly, the  
22 solenoidal coil array may have cascaded 10' (FIG. 4), overlapped  
23 10' (FIG. 5) or sandwiched 10'' (FIG. 6) configurations depending  
24 on if the second solenoidal coil 14 is outside the UCR solenoidal  
25 coil 12, overlapped with section B of the UCR coil 12 or inside  
26 the UCR coil 12, respectively. In any case, the solenoidal array  
27 is UCR-based and is conceptually different from and more advanced  
28 than the previous "sandwiched solenoidal array" due to the  
29 advantages associated with the quasi-one-peak profile feature of  
30 the UCR design.

31 In a UCR-based solenoidal array, each coil is subjected to  
32 noise pickup from a smaller region just like other types of array

1 coil. The array coil advantages in terms of signal to noise ratio  
2 and field of view improvement applies to the UCR solenoidal array  
3 as disclosed in this invention.

4 A prototype UCR solenoidal array was built to prove the  
5 concept. The prototype solenoidal array coil included a UCR  
6 solenoidal coil and a 2-turn solenoidal coil. The solenoidal coil  
7 array was built for a 0.3T Hitachi Airis II imaging system at the  
8 resonance frequency of 12.687 MHz.

9 The coil traces were made of 0.2 mm thick, 10 mm wide,  
10 copper strips wound on a 267 mm diameter acrylic tube. The two  
11 solenoidal coils of the array were in overlapped configuration,  
12 meaning that the 2-turn solenoidal coil 14 overlaps with the  
13 section B of the UCR coil 12. The geometric parameters are shown  
14 in FIG. 7. The coils 12, 14 are shown on separate axes for ease  
15 of understanding.

16 The two solenoidal coils 12, 14 are inherently decoupled.  
17 Excellent isolation was achieved between the two coils without  
18 any additional isolation circuitry. The transmission parameter  
19 S21 is -28dB at resonance frequency, as shown in FIG. 8.

20 The B1 field along the axial direction was measured for each  
21 solenoidal coil alone, with the other coil active. The results  
22 are shown in FIG. 9. Also shown is the combined solenoidal array  
23 B1 profile. The UCR solenoidal coil 12 has a quasi-one-peak  
24 profile with a null-B1 point residing at about the middle of the  
25 coil 14 profile. The summation of individual profiles gives a  
26 nice total array profile without artifacts.

27 The array coil of the invention need not be just solenoidal  
28 coils. For example, an orthogonal coil element, such as a saddle  
29 coil, may be added to form a quadrature pair with each solenoidal  
30 coil. Therefore, a two-solenoidal coil array can be easily  
31 developed to be a two-quadrature-pair solenoidal array coil to  
32 take advantage of quadrature effect in signal to noise ratio

1 improvement.

2       It should be evident that this disclosure is by way of  
3 example and that various changes may be made by adding, modifying  
4 or eliminating details without departing from the fair scope of  
5 the teaching contained in this disclosure. The invention is  
6 therefore not limited to particular details of this disclosure  
7 except to the extent that the following claims are necessarily so  
8 limited.

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